Evaluation of High Temperature Superconducting (HTS) Coil Performance Gain for 3T Small Animal MR Imaging

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Introduction It has been widely reported that substantial signal to noise ratio (SNR) gain can be achieved through the use of the high-temperature superconducting (HTS) surface coil. Due to its complex and expensive design and construction, computational simulation is an essential design tool [1]. However estimating the likely performance gain of the HTS coil where accurate modelling of important parameters such as loaded Q factor and SNR gain in MR imaging experiments has not yet been reported. Firstly this work reports an FDTD mesh based numerical mouse model which will accurately simulate a realistic loading effect on the HTS coil at 3T. Secondly, techniques are developed to assess the high Q factor and SNR gain, and comparisons are made with that of the room temperature copper coil. Finally MR imaging experiments performed on room temperature copper coils to further validate the accuracy of the developed method.

Method and Results *Design of the HTS coil:* In order to avoid the resistance introduced by the normal chip capacitors and connections to the small size HTS coil, a double-sided self-resonator design is proposed for 3T MR imaging. This design was simulated using the FDTD method (xFDTD, Remcom. Inc, PA) [2]. The geometry is shown in Fig. 1, where two three-turn square HTS coils with an inner diameter of 10.8 mm are deposited on the top and bottom of a 0.43 mm thick sapphire substrate ($\varepsilon_r = 9.4$). An additional prototype coil was initially fabricated by using copper material to validate the accuracy of the numerical simulation. Then the original HTS coil design was revised by incorporating an 8% deviation (due to true substrate permittivity) between the simulation and experiment. The final HTS coil achieves a desired resonant frequency with unloaded Q factor around 12000, which is 100 times higher than that of the room temperature copper coil.

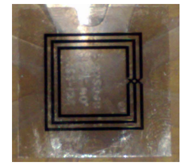

Fig. 1: Double sided HTS coil with opposite gap opening

FDTD mesh based numerical mouse model: The anatomically detailed FDTD grid of a 28g nude normal male mouse model is created by first retrieving pixels from 206 coronal images of a "digimouse" atlas (http://neuroimage.usc.edu/Digimouse.html) with 0.1 mm resolution, and transforming these segmented images into a 3D grid of Yee cell cubes. A computer program was written to perform the transformation and check the continuity of skin on the outer surface of the model. In order to save simulation time, the fine mesh was then rescaled to coarser 1 mm mesh, and several error-correction procedures were performed to examine mesh edge discontinuities. The mesh consists of twenty tissue types, and electrical parameters of these tissues at 128 MHz are used in the EM simulation (http://niremf.ifac.cnr.it/tissprop/htmlclie/htmlclie.htm#atsftag). The numerical geometry is shown in Fig. 2 where the designed HTS probe is located 1 cm above the mouse model to accommodate the cryogenic thermal insulation. The simulation is driven by a drive loop placed 4 mm above the RF probe to simulate the inductive coupling loop as utilised in the real MR experiment. From the simulation, the resonance frequency of the loaded HTS coil is shifted down by 0.126 MHz compared with that of unloaded case as shown in Table 1. Fig. 3 shows the magnitude of the magnetic field at a coronal plane 1.6 cm away from the coil. The magnetic field has a homogeneous field-of-view within the heart region of the mouse model. The coil and cryostat system has been designed and optimised for rodent cardiac studies.

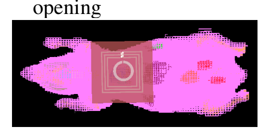

Fig. 2: FDTD simulation geometry of a loaded HTS coil

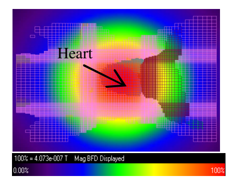

Fig. 3: magnetic field inside the mouse model

Simulation of the High Q factor of the loaded HTS coil: The widely used method to calculate the Q factor in an EM simulation is to Fourier Transform the time course of the electric field response of the R.F. probe [3]. However this method requires long run time to achieve a high frequency resolution $\Delta f \approx 300kHz$ in order to accurately predict a high Q factor associated with the HTS coil (which is normally above 10000). Therefore another method to calculate the Q factor is proposed: The inductive coupling probe is excited by a sinusoidal wave at ω_o =128 MHz. Once steady state has been reached, the Q of the loaded HTS coil can be calculated as: $Q_{loaded} = \omega_o W_s / P_L$ [1]. Where the time-average stored energy W_S of the RF probe is expressed as $W_s = \frac{1}{4}\int_t^{t+T}\iiint_V \varepsilon|E|^2 + \mu|H|^2\, dxdydzdt$ and the total power loss P_L is calculated as: $P_L = \frac{\sigma}{2}\int_t^{t+T}\iiint_V |E|^2\, dxdydzdt$. The integrations are performed at each time step by summing the magnetic and electric field values in the mesh weighted by the associated cell size, and averaged over a complete cycle of the input sinusoidal wave. Table 1 shows Q factors and resonant frequency shifts of the loaded HTS probe and its copper mimic. Compared with the unloaded Q factor around 12000, the Q factor for the loaded HTS coil has been reduced by 50%, to around 6000, which means the HTS coil quality is largely determined by the noise induced by sample.

Coil Type	$Q_{unloaded}$	Q_{Loaded}	Frequency Shift When Loaded
HTS Coil(77 K)	12703	6367	0.126 MHz
Copper Mimic (77 K)	280	241	negligible
Copper Mimic (293K)	75	70	negligible

Table 1: Q factors and resonant frequency shifts of the loaded HTS probe and its copper mimic at 300 K and 77 K.

SNR gain by using the HTS coil for small rodent imaging: The SNR distributions of coils made from different materials loaded with the numerical mouse model were simulated by calculating SNR $\propto (B_1/I)/\sqrt{R_{eq}}$, where B_1 is the magnitude of the B_1 field, I is the input current, R_{eq} is the total equivalent resistance composed of the sum of sample resistance and coil resistance. The SNR distributions of the HTS coil, copper coil at 293 K and 77 K at a point 2 cm from the coil inside the mouse sample against different coil sizes are shown in Fig. 4. The data suggest that using the 1 cm HTS coil should improve the SNR by a factor of 3 over that of a copper coil at 293 K. In order to validate the SNR simulation, we built three copper loops at 293 K with different diameters as shown Figure 4(a) and tested their SNR performances by imaging a plastic tube filled with physiological saline solution. The SNR performances of these three coils agree well with the simulated copper coil performances.

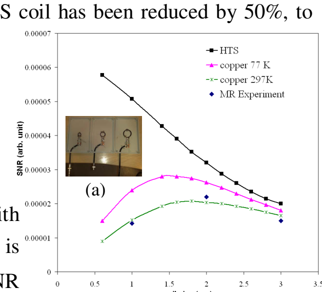

Fig. 4: SNR distributions of different coils

Conclusion and Discussion In this work, the sensitivity of the HTS coil is evaluated by using the FDTD simulation with the aid of an anatomical mouse model. Comparisons of simulations with experimental data yielding Q factors and SNR measurements of copper coils have demonstrated a good agreement with the simulation prediction. Validation of SNR and Q using the HTS coil in an imaging system is currently awaiting construction of a cryostat suitable for rodent imaging.

References:

[1] J. Nouls et.al., Design of a Superconducting Volume Coil for Magnetic Resonance Microscopy of The Mouse Brain, Journal of Magnetic Resonance 191 (2008), 231-238. [2] B. Hu et.al., Electromagnetic Simulation Guided Design of an Optimised Self-Resonant HTS Receive Coil for 3T MRI, British Chapter ISMRM 13, (2007). [3] T. Ibrahim, Design of Radiofrequency Coils for Magnetic Resonance Imaging Application: A computational Electromagnetic Approach, PhD Thesis, (2003).